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Welter, T.G.; Bobbert, M.F.

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Initial Arm Muscle Activation in a Planar Ballistic Arm Movement With Varying External Force Directions: A Simulation Study

Tom G. Welter and Maarten F. Bobbert

It has been shown in previous research that the initial phase of EMG for a punching movement remained almost unchanged regardless of whether an external force was applied to the arm. The purpose of the present study was to explain this finding with the help of simulations. A two-dimensional model of the arm actuated by 6 Hill-type muscles was used to simulate a punching movement in the horizontal plane from a prescribed starting position with 90° elbow flexion. Input to the model was the stimulation of the muscles, and output were, among others, muscle forces and segmental accelerations. A genetic algorithm was used to determine the muscle onset times that minimized movement duration and targeting error. In a subsequent forward simulation, the optimized muscle onset times for an unloaded punching movement were superimposed on the isometric stimulation necessary to hold the arm in the starting position while an external force was applied to the arm. The resulting movement was only slightly different from the unloaded movement. It appeared that because of the low level of isometric muscle force prior to the movement, and the high level of stimulation during the movement, muscle force was increased at a rate that was almost independent of the prior force level. These results confirmed the suggestion that the initial phase of EMG in ballistic movements is more related to the rate of change of force than to the absolute force level. It is hypothesized that this may simplify the task of the nervous system in the choice of initial muscle activity in ballistic arm movements because no adjustments to varying external forces are required.

Key Words: movement simulation, EMG, ballistic movements, activation dynamics, motor control, biomechanics

Introduction

Researchers typically measure EMG in order to study the rules by which the nervous system selects muscle activity. These studies attempt to relate EMG to muscle forces and joint torques, or to the dynamics of the movement. Although different

The authors are with the Institute for Fundamental and Clinical Human Movement Sciences in the Faculty of Human Movement Sciences at Vrije Universiteit Amsterdam, van der Boechorststraat 9, 1081 BT Amsterdam.

relations between EMG and movement dynamics have been proposed by various authors (Flanders, Pellegrini, & Geisler, 1996; Gottlieb, 1996; Hong, Corcos, & Gottlieb, 1994; Prilutsky, 2000; van Bolhuis, Gielen, & van Ingen Schenau, 1998; van Ingen Schenau, 1989), they all predict EMG changes for a given movement when external or inertial forces are varied. Recently, however, we found that the integrated initial 100 ms of the rectified EMG burst (*initial EMG*) for ballistic movements (200–300 ms) in a given direction remained almost unchanged regardless of whether an external force was applied to the arm. This was true for all movement directions and for all muscles relevant to a specific movement direction (Welter & Bobbert, 2002). Subjects were instructed to punch a cushion as fast as possible, starting at a position of 90° arm abduction and elbow flexion (Figure 1). Punches were made in the horizontal plane, either with or without an external force (25 N) applied at the wrist. When a force was present, counteracting joint torques had to be generated by muscles prior to and during the movement. The range of isometric torques associated with different force conditions was of the same magnitude as the range of torque variations that occurred during the movement. Despite the different isometric muscle activity (*isometric EMG*) required to generate joint torques ranging from flexor to extensor prior to the movement, only small amplitude changes of the initial EMG burst appeared to be necessary to produce similar results in terms of movement direction and velocity (Figure 3, first column).

This observation is in contrast to the assumption that changes in external forces require adaptations in muscle stimulation (EMG). As a possible explanation for the small EMG changes, we suggested that EMG in fast movements is dominated by stimulation necessary for the fast increase of muscle force (Welter &

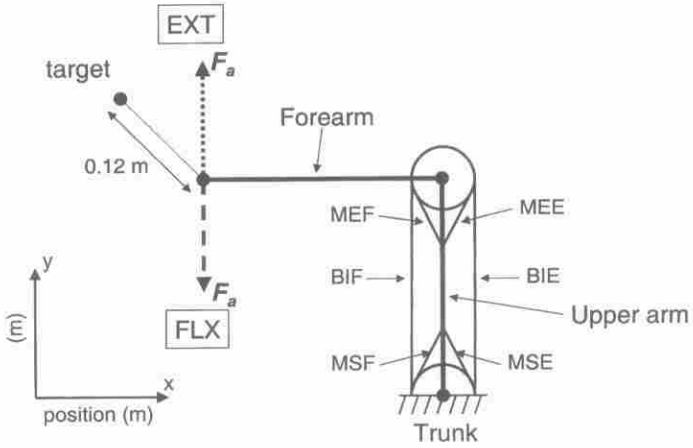

Figure 1 — Schematic of the arm model and task conditions. The arm model moves in the horizontal plane. Starting position was with the elbow flexed at 90°. Movements towards the target were executed without load on the distal end of the arm and with an external load requiring a compensating force F_a exerted by the arm, generated by elbow extensor torque (EXT condition, dotted arrow) or flexor torque (FLX condition, dashed arrow). The origin of the coordinate axes is located in the shoulder joint.

Bobbert, 2002). This would be in line with the observation that initial EMG amplitude was very high compared to the isometric EMG amplitude, which could not be explained by the force–velocity relationship. In ballistic movements, activation dynamics (i.e., the process of cross bridge formation in response to neural stimulation of the muscle; the process of building up the active state) becomes increasingly important because it determines the speed at which force can be increased. It may be that the nervous system uses "extra" stimulation to increase this speed. EMG burst amplitude would then mainly be an indication of the rate of increase of muscle force rather than the absolute force level.

If this explanation is correct, the use of one muscle stimulation pattern should produce a good, albeit not perfect, initial direction of a ballistic movement regardless of the external force that is applied to the arm, in a situation where the external force is compensated in advance by isometric muscle activity. The aim of the present study was to investigate this suggestion using a two-dimensional model of the arm, actuated by six Hill-type muscles. Stimulation onset times for a punch in a given movement were determined through optimization. Subsequently, these onset times were used to simulate punches when external forces, comparable to those in our previous experiment, were applied to the arm.

Methods

General Outline

An arm model (Figure 1) was used to simulate punches in the horizontal plane to a given target. Simulations were performed first without external force, requiring no joint torques prior to the movement (unloaded condition), and then with forces (30 N) applied to the tip of the arm, requiring counteracting elbow extensor torque (EXT: extensor condition) or flexor torque (FLX: flexor condition; Figure 1). These force conditions were comparable to those in our previous experiment (Welter & Bobbert, 2002). The target direction was selected because (a) it required considerable elbow joint torques in the same order as those required to counteract the external force; (b) if our explanation is not correct, deviations from the intended movement path would be expected when the same stimulation pattern is used in different force conditions because of the angle between movement and force direction; and (c) the data could be compared to experimental data from the same movement direction. The target distance was 0.12 m. The model needed approximately 120 ms to reach the target, which enabled a comparison between the initial 100 ms of simulated movement and the experimental data. The simulations started from a fixed position and ended when the tip of the forearm reached the target position. For all force conditions, a set of stimulation onset times per muscle (timing pattern) was determined through optimization.

Computer Model of the Arm

The arm model consisted of two segments in the horizontal plane representing the forearm (plus hand) and the upper arm, connected to an immobile "trunk" (Figure 1). Segments were interconnected by two frictionless joints representing the elbow and shoulder. Segment parameters were taken from Winter (1979) and are presented in Table 1. Forearm length was taken without the hand because in the experiments, force was applied at the wrist.

Table 1 Segment Parameters

Segment	Length (m)	Mass (kg)	CoG (m)	J (kg · m^2)
Forearm + hand	0.263	1.65	0.179	0.025
Upper arm	0.335	2.1	0.146	0.0244

Note. Segment parameters according to Winter (1979). CoG: distance of center of gravity relative to the proximal joint. J: moment of inertia relative to the center of gravity. Forearm length was without hand, but the other three parameters included the hand.

Six lumped muscles spanned the elbow and shoulder joint: a mono-articular shoulder flexor (MSF) and extensor (MSE), a mono-articular elbow flexor (MEF) and extensor (MEE), and a bi-articular shoulder and elbow flexor (BIF) and extensor (BIE). The lumping of muscles was based on a set of 19 relevant muscles that were lumped into six muscles for an arm model by Nijhof and Kouwenhoven (2000). Maximum isometric force for each lumped muscle was calculated using joint torque data and physiological cross-sectional area (PCSA) data from Nijhof and Kouwenhoven (2000) in combination with moment arm data. We assumed the same force per unit cross-sectional area for all muscles. A Hill-type muscle model was used that has been described in detail elsewhere (van Soest & Bobbert, 1993). Parameters of the force–velocity relationship were taken from data of human calf muscle (van Zandwijk, Bobbert, Harlaar, & Hof, 1998; see also Welter et al., 2000). The concentric force–velocity characteristic is based on the classic description by Hill (1938). The shape of this curve is determined by the dimensionless parameter a_{rel} and the parameter b_{rel} (s^{-1}), being a/F_{max} and $b/L_{ce,opt}$ ($L_{ce,opt}$: optimum muscle length), respectively, where the parameters a (N) and b (m/s) are the usual parameters in the Hill equation. Values for a_{rel} and b_{rel} were 0.21 and 2.01, respectively. Optimum lengths of the contractile elements, lengths of the series elastic components (tendon lengths), and moment arm data were obtained by averaging data from the literature (C.W. Spoor, personal communications; Murray, Buchanan, & Delp, 2000; Murray, Delp, & Buchanan, 1995; Table 2). Compliance of the series elastic components was set at 4% at maximal isometric force.

Input to the model was the stimulation (*stim*) to the muscles, a number between 0 and 1 being a one-dimensional representation of recruitment and firing rate of the motoneurons. The active state (fraction of the available number of binding sites for cross bridge formation that is attached) was related to *stim* by first-order dynamics described by He (He, Williams, & Loeb, 1991). Since we wanted to simulate maximally fast (ballistic) movements, and we speculated that high EMG (*stim*) levels were mainly responsible for fast rates of force change, *stim* switched to 1 at stimulation onset. Initial values of *stim,* necessary for static equilibrium in conditions where an external force was present, were calculated by first minimizing muscle stress cubed (Crowninshield & Brand, 1981; Prilutsky, 2000) that could still provide the necessary joint torques, and then calculating the *stim* level that resulted in the calculated force for each muscle. Initial *stim* levels ranged between 0.005 for "silent" muscles and 0.21 for active muscles. The dynamic

Table 2 Muscle Parameters

Muscle	MSF	MSE	MEF	MEE	BIF	BIE
Lce_{opt} (m)	0.1337[c]	0.1397[c]	0.0850[a]	0.093[a]	0.0905[a]	0.127[a]
F_{isom} (N)	1768[b]	1573[b]	1648[b]	1332[b]	464[b]	492[b]
Lse (m)	0.039[c]	0.066[c]	0.141[a]	0.187[a]	0.206[a]	0.217[a]
A0	0.19326796 (s)	0.27293204 (s)	0.24265460 (e)	0.25141039 (e)	0.2509993(s), 0.1(e)	0.20945617(s), 0.1(e)
A1	–0.03	0.03	–0.01	0.03	–0.03(s), –0.015(e)	0.03(s), 0.03(e)
A2	0	0	–0.0043	–0.0024	0(s), –0.0043(e)	0(s), –0.0024(e)

Note. Lce_{opt}: optimal length of contractile element; F_{isom}: maximum isometric force, estimated using moment arms, estimated maximum isometric joint torques, and PCSA data; Lse: length of series elastic component (tendon stretch was 4% at F_{isom}); A0, A1, A2: coefficients for the relation between muscle-tendon complex length and joint angle ($L_{mtc} = A0 + A1 \cdot angle + A2 \cdot angle^2$); e = elbow; s = shoulder. Moment arms around the elbow were based on data from Murray et al. (1995); moment arms around the shoulder were estimated.

[a]Murray et al. (2000); [b]Calculated from data of Nijhof and Kouwenhoven (2000); [c]Calculated from data of C.W. Spoor (personal communications).

optimization problem consisted of finding the muscle onset times (timing pattern) that minimized the sum of (a) the time necessary to displace the hand over a distance of 12 cm (in seconds, multiplied by 10), and (b) the distance between the hand position at that instant and the target position (in cm, squared). A genetic algorithm was used for the minimization (van Soest & Casius, 2000).

The model output was compared to data from our previous experiments (Welter & Bobbert, 2002). Setting the time constant of the activation dynamics to 15 ms (Winters & Stark, 1987) resulted in joint torques that were higher and much faster than the experimental data. To match the rate of torque change (calculated as the rate of change between 10% and 90% of the maximum torque in the unloaded condition) of the model to the experimental data (Figure 2 D), the time constant of activation dynamics was set at 137 ms. The time constant for deactivation was not important because muscles were not switched off in the simulations.

Results

Comparison Between Simulated and Experimental Data

Figure 2 shows "hand" path, velocity, and acceleration together with elbow joint torques of the unloaded punch for the model (solid lines) and the experiment (dashed lines, mean of 10 subjects). The model data appeared to be advanced by approximately 25 ms compared to the experimental data. This seems mainly due to a

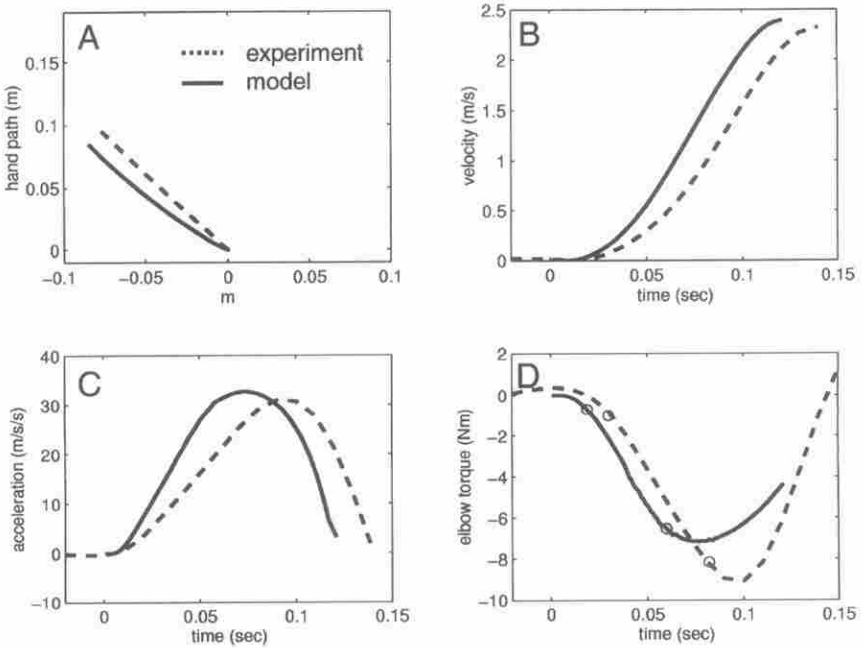

Figure 2 — Comparison between model (solid line) and experimental data (dashed line; Welter & Bobbert, 2001). t = 0: the instant that the first muscle activity was detected (experiment) or the start of the simulation (model). Circles in D indicate 10% and 90% of the maximum torque. The coordinate axes in A are the same as in Figure 1, with the origin shifted to the start position of the hand.

faster initial rise of the simulated joint torques that was not observed in the experiments. However, the slopes and amplitudes of the joint torque, and the hand acceleration and velocity, were comparable. Therefore, no further attempt, other than for the time constant of activation dynamics mentioned above, was made to obtain closer agreement with experimental data by means of tuning the model parameters.

One Timing Pattern Can Be Used Regardless of External Force

Figure 3 shows that identical stimulation patterns can be used to generate a punching movement in a given direction regardless of the external force. The first column shows the experimental data that motivated the current simulations. EMG data (panel B) are from the bi-articular m. triceps caput longum. The *stim* patterns in panels F and J are from the corresponding BIE. (The *stim* pattern for MEE is shown in Figure 4.)

The second column (panels E–H) shows data from the optimization in the unloaded condition, together with the effect of the "unloaded" onset times superimposed on the isometric *stim* levels in EXT and FLX. It can be seen that only in EXT, the arm tip missed the target by only a few centimeters despite differences in stimulation level and joint torque prior to the movement in EXT and FLX (panels F and G, $t = 0$). The error is not larger than the difference between the average movement directions in the experimental data (compare panels E and A). The EXT condition showed a deviation to the left with respect to the target. In contrast, the experimental data showed a deviation to the right for the EXT condition. (See the discussion section for possible explanation.)

The third column of Figure 3 (panels I–L) shows data from optimized onset times in all force conditions. Small adjustments in the onset timing of BIE were necessary to reach the target (compare panels F and J). However, there were no major changes in torque histories (panel K) or velocity histories (panel L) compared to column 2.

It can be seen that in the experiment, the elbow torque curves for the different conditions were similar except for the offset at $t = 0$ that depended on the external force condition (panel C). Despite this difference in torque offset, EMG burst amplitude remained almost unchanged in the three conditions (panel B). In the simulations, torque curves displayed variations in amplitude and offset (panel G) comparable to the experimental data, but now as a result of identical *stim* levels after muscle onset (panel F). Shoulder torque curves displayed the same characteristics as in the elbow—that is, the curves in the three conditions were similar except for the offset at $t = 0$. Therefore, these data are not presented.

Note that the elbow torque variations in the unloaded condition were somewhat larger than in FLX and EXT (panels G and K), resulting in a slightly higher movement velocity in the unloaded condition (panels H and L).

Low Active State Levels and High Stimulation Lead to Comparable Rates of Force Increase

The data in Figure 4 are used to make clear how it is possible that the same elbow extensor stimulation pattern (panel A; see also Figure 3F) can be used for movements with different external force conditions. The figure shows data for MEE, which is a larger contributor to elbow torque than BIE.

At the instant that stimulation is switched on (panel A), muscle force started to increase in all conditions (panel C) at a rate largely determined by the rate of

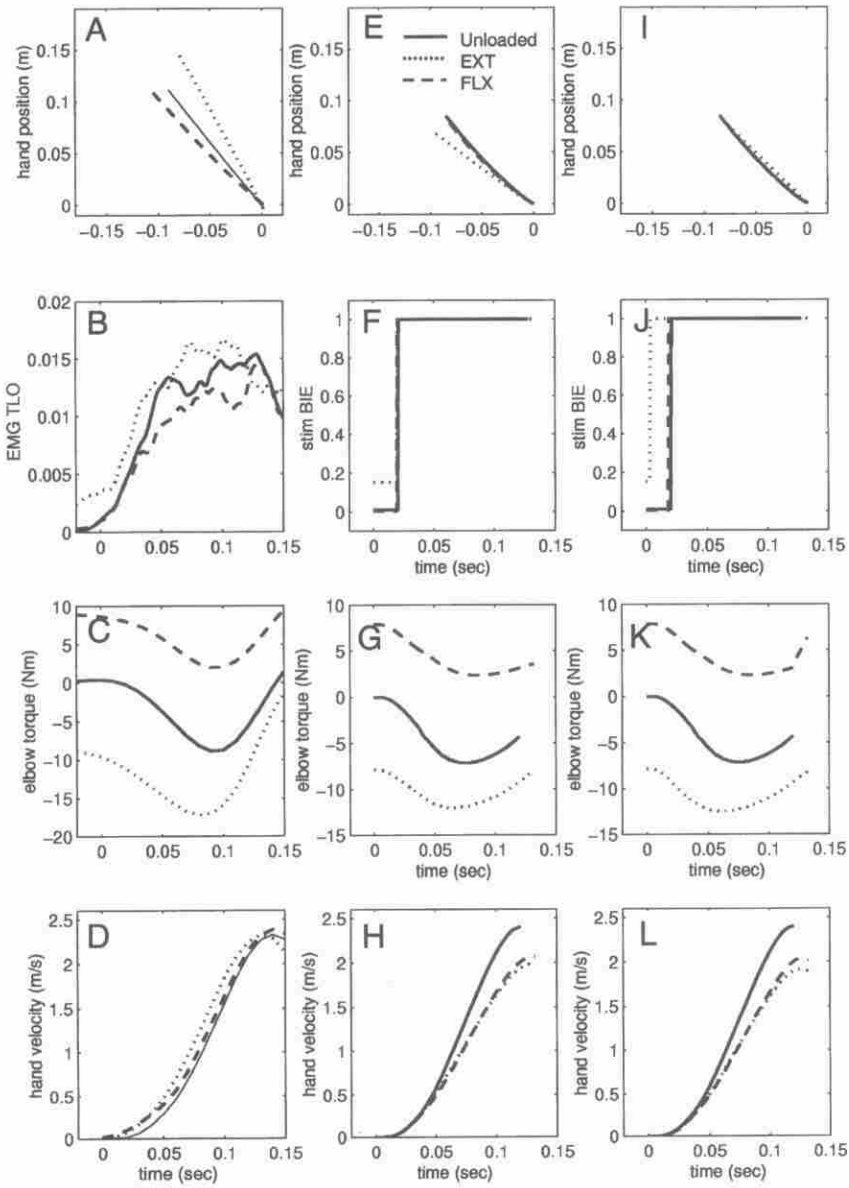

Figure 3 — Experimental and model data for unloaded (solid lines) and external force conditions (EXT: dotted, FLX: dashed). Panels A–D: experimental data. Note the similar EMG curves (panel B) despite the difference in joint torque (panel C). The small difference between EMG curves in different load conditions is typical for the EMG differences of other muscles. TLO: m. triceps caput longum. Panels E–H: optimized onset timing for the unloaded condition applied to both loaded conditions. Note the small errors in hand paths (panel E) despite the identical onset timing (panel F) and the similar torque variations in all force conditions. Panels I–L: onset timing optimized for each external force condition. The coordinate axes in A, E, and I are the same as in Figure 2.

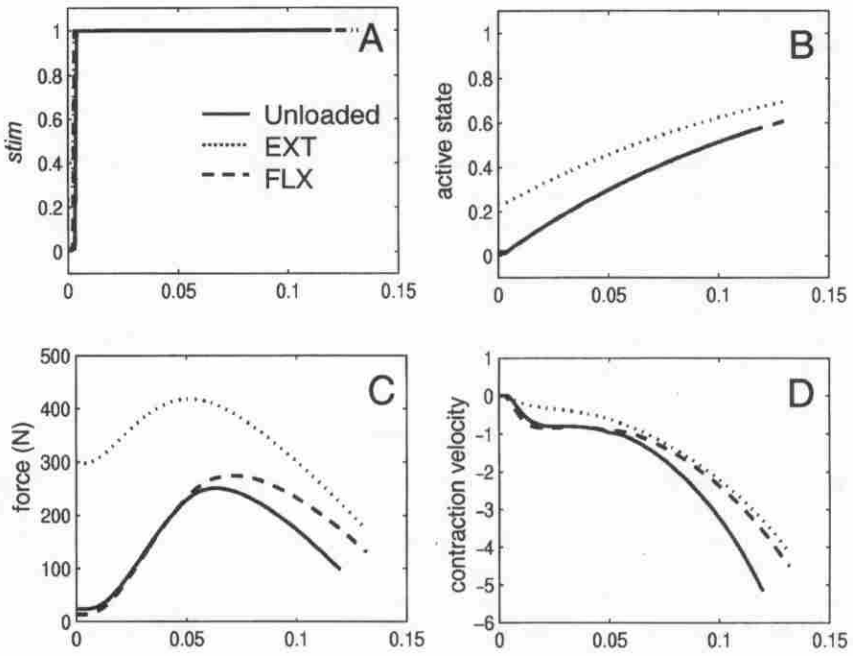

Figure 4 — A: *stim* level, B: active state level, C: force level, D: contraction velocity (expressed in optimum contractile element length / s) for the mono-articular elbow extensor (MEE), optimized for the three force conditions. Note the different force and active state levels despite the identical *stim* levels.

change in active state (panel B). In the first 50 ms, the force–length and force–velocity relations had little influence on muscle force because contraction velocity was low (panel D). After 50 ms, muscle force started to decrease despite the still increasing active state because of the increased contraction velocity. The reason that active state, and therefore muscle force and joint torque, rose at comparable rates in the different force conditions is that the active state levels prior to the movement were relatively low compared to the "target" active state level that was set when the muscle was switched on. The resultant effects of prior activity level on the rate of force increase are small. The slightly reduced rate of change of active state in EXT that resulted from the prior *stim* level (panel A) led to a smaller force increase (panel C) than in the unloaded and FLX condition and, eventually, to a somewhat lower peak velocity than in the unloaded condition (Figure 3H).

The relatively low active state level prior to the movement is important in the explanation given above. It is illustrated by the low isometric force level in MEE of 300 N (Figure 4C) compared to the maximum of 1332 N (Table 2), and by the low isometric elbow extensor torque of 8 Nm (Figure 3G, K) compared to the maximum of 41 Nm (Nijhof & Kouwenhoven, 2000) that was used to calculate the maximum isometric force for each muscle. Although the active state level is relatively low, this does not mean it is not important. When the prior muscle activity was not present in the loaded conditions—that is, when in simulations the external

force was applied at the instant that the movement started—the target was not reached at all, and the arm was pulled in the direction of the external force.

Note that elbow torque (Figure 3G) and muscle force (panel 4C) were quite different between the three conditions considering that identical *stim* levels were used. It can be concluded that the *stim* level after stimulation onset mainly affected the rate of change of muscle force, and did not correlate with absolute muscle force. Active state did not reach the maximum level due to the relatively slow activation dynamics, even in muscles that were activated at the start of the simulation (panel 4B).

The explanation above for the small influence of external force on the resulting movement direction not only held for the target direction presented; additional simulations showed that in other movement directions, the error was also small when *stim* was not adjusted (Figure 5).

Discussion

The simulations show that only one stimulation pattern with fixed level and timing can be used to generate punches in a given direction, assuming that any external force is compensated in advance by isometric muscle activity. This is because the active state level at stimulation onset is low compared to the target level for both "silent" and active muscles. The results support our earlier suggestion that the initial EMG in ballistic movements is related to the rapid change in muscle force and has no relation to the absolute force level. However, before drawing a definite conclusion and speculating on the implications for motor control, we will first address the validity of the model simulations.

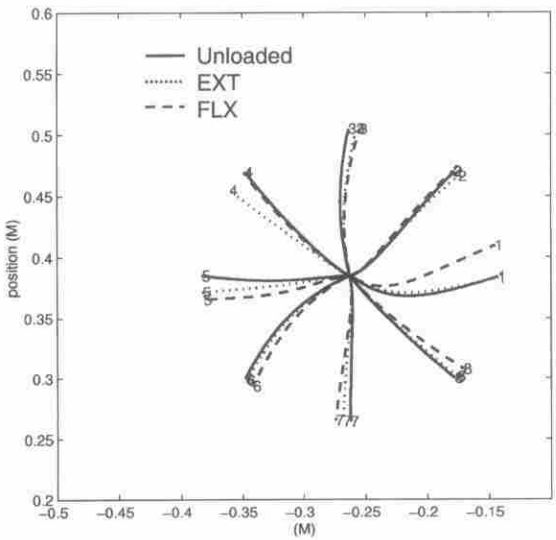

Figure 5 — Movement trajectories of the tip of the arm in 8 different movement directions. The optimized onset timing for a particular direction from the unloaded condition was used in both loaded conditions. Trajectory number 4 is the same as in Figure 3E. Solid line: unloaded condition; dotted line: EXT condition; dashed line: FLX condition. Coordinates are the same as in Figure 1.

Validity of the Model Simulations

When an arm model is used in simulations, it must reproduce the behavior of a real arm in order to avoid conclusions that are only valid for the model and not for the real system. In the present model, all data were taken from literature, and no special tuning of model parameters was performed except for the time constant of the activation dynamics. The fact that this straightforward approach resulted in behavior that was comparable to experimental data (Figure 3) is an indication that the model captured salient properties of the human arm.

It could perhaps be argued that the robustness of onset timing for different external force conditions was an artifact caused by the "slow" activation dynamics that were used. Therefore, simulations were repeated with infinitely fast activation dynamics—that is, active state was equal to *stim,* which switched from 0 to 1 instantaneously. It appeared that the speed of activation dynamics did not alter the conclusion that one stimulation pattern can be used to generate punches in a given direction under different external force conditions. A high speed of activation dynamics only increased movement dynamics to unrealistic values.

Furthermore, the validity of the model simulations may be questioned because of the simple model of activation dynamics (He et al., 1991) that was used. Therefore, we also performed simulations with the more detailed model of activation dynamics by Hatze (1981). It appeared that, although the timing difference between model and experiment (Figure 2) was avoided, muscles had to be made extremely slow—and even then *stim* could not be set higher than 0.5—in order to get realistic movements. However, the conclusions were not different from those presented here. Therefore, we only presented data from the simplest model.

The movement velocity of the model was lower in the loaded conditions than in the unloaded condition, whereas in the experiment, the peak velocity was constant regardless of load (Welter & Bobbert, 2002). This may be related to the use of fixed *stim* levels used in the model. First, in the experiment, the EMG amplitude that was reached in muscles that were active prior to the movement in the loaded condition was slightly higher than in the unloaded condition (Figure 3B). This was especially so after the period of 100 ms (see Figure 3D in Welter & Bobbert, 2002). In the model simulation of the EXT condition, the fixed *stim* level in BIE and MEE could not compensate for the slightly reduced rate of change in active state that was found. Second, antagonist muscles often showed decreased EMG in the experiments, sometimes even before the onset of agonist activity. In the model, antagonist muscles could not be switched off. In FLX, the length increase of the active flexor muscles during elbow extension caused an increase in flexor muscle force, which reduced the net changes in extensor joint torque and consequently decreased movement velocity.

These fixed *stim* levels could also be the reason that movement paths of the model differed from the experimental paths (Figure 3A, E). An extra high MSE stimulation level in the model (comparable to the higher EMG found in the experiments) could have increased the extensor torque in the shoulder, preventing the movement deviation to the left in the EXT condition.

Implications for Motor Control

The present findings indicate a simplification in the choice of initial muscle stimulation pattern in ballistic movements. It seems that one open-loop controlled

stimulation pattern that is specific for a certain movement direction can simply be superimposed on the current muscle stimulation and still result in a roughly correct movement execution during the first 100 ms. This agrees with the suggestion that motor planning has multiple stages (Koshland & Hasan, 1994; Koshland, Marasli, & Arabyan, 1999; Sainburg, Ghez, & Kalakanis, 1999). The first stage would then consist of a simple open-loop control because of the limited nerve conduction velocity. Subsequent stages would be controlled by increasing usage of feedback.

Simulations cannot tell us the rules that may be used by the nervous system in the selection of muscle activity. However, based on the present results, it can be concluded that for the initiation of fast movements, these rules do not have to incorporate the knowledge of present and future torque values. For example, Karst and Hasan (1991) proposed that the initial choice of muscles might be based on the direction of the movement target relative to the forearm, without regard to necessary joint torques. The present data suggest that such a rule might also work when external forces are acting on the arm or when the arm is rotated with respect to gravity.

As mentioned in the introduction, many authors attempt to relate EMG to muscle forces and joint torques. Obviously, the finding that the same stimulation pattern can be used regardless of the external force direction means that these attempts are doomed to fail. It appears that for the initiation of ballistic movements, EMG amplitude must be related to *changes* in muscle forces, available through changes in joint torques.

Conclusions

Identical stimulation patterns can be used to start punches in a certain direction, regardless of the external force on the wrist, assuming that any external force is compensated in advance. The resulting differences in movement execution will be small during the first 100 ms. This is because the active state level in silent and active muscles prior to the movement is low compared to the high target active state level that is set when the muscle is switched on. As a result, the rate of change in muscle force is almost the same in "silent" and active muscles. This explains why in punching movements, EMG levels seem to be independent of external force levels.

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